RADIATION MEASUREMENT WITHIN THE HUMAN BODY

Cross-Reference to Related Patent Application

This application is a continuation-in-part of application serial number [Attorney Docket: tes1] filed on January 28, 2002 by the inventor hereof and entitled "RADIATION MEASUREMENT WITHIN THE HUMAN BODY", the specification of which is incorporated by reference herein.

Background of the Invention

1. Field of the Invention

The present invention pertains to the field of radiation measurement within the human body. More particularly, the present invention pertains to the field of in-vivo dosimetry using a single scintillating fiber inserted into the human body for accurate radiation measurement.

2. Description of the Related Art

Radiation is often used in the treatment of human health problems. These problems, generally occurring "in-vivo", include cancer and, more recently, the re-closing of arteries after balloon angioplasty. Knowledge of radiation levels within the body, important for the success of such treatment, is usually pursued by calculation or modeling because direct measurement of internal body radiation is often too difficult to carry out.

In the field of radiation measurement within the human body, the following related art documents are known (all of which are incorporated herein by reference): U.S. Patent 4,932,412 entitled "Intraoperative and Endoscopic Tumor Detection and Therapy"; U.S. Nuclear Regulatory Commission Report NUREG/CR-5223 entitled "Scintillating Fiber Detector for In-Vivo Endoscopic Internal Dosimetry", published October 1988; Phys. Med. Biol., 1992, Vol. 37 No. 10, pp.1883-1900 entitled "Water-equivalent plastic scintillation detectors for high-energy beam dosimetry: I. Physical characteristics and theoretical considerations"; U.S. Patent 5,704,890 entitled "Real Time Sensor for Therapeutic Radiation Delivery"; Japanese Published Unexamined Patent Application 10-213663 entitled "Local Dosimeter"; U.S. Patent 5,880,475 entitled "Scintillation Fiber Type Radiation Detector"; U.S. Patent 5,905,263 entitled "Depth Dose Measuring Device"; U.S. Patent 6,151,769 entitled "Method of Making a Scintillator Waveguide"; and Japanese Published Unexamined Patent Application 2001-56381 entitled "Local Radiation Amount Measuring Device and Medical Device Equipped Therewith".

The systems noted above require special corrections or fiber assemblies to remove errors arising from Cerenkov radiation, which generally complicates the use of scintillating fibers for therapeutic radiation. Cerenkov radiation is the radiation that results when a charged particle, moving in some medium, travels faster than light does in that medium. (The charged particle can be introduced 'on its own', or can be an electron kicked out of an atom by an entering photon = gamma ray or X-ray.) The speed of

light in a medium is given by its index of refraction. The speed of a particle is given by its mass and its energy. The index of refraction in most scintillating fiber materials is 1.6. Thus, an electron with energy more than about 0.14 MeV introduced on its own, or kicked out of an atom by an entering photon (= gamma ray) of energy more than about 0.28 MeV, will be travelling fast enough to generate Cerenkov radiation. Most High Dose Rate radioisotopes (e.g. Ir-192 and P-32, which are introduced into the body via a catheter or the like during HDR afterloader therapy), and all 'external X-Ray beam' sources, will have enough energy to trigger Cerenkov radiation. Low Dose Rate radioisotopes (e.g. I-125, Pd-103) used in medical implant seed therapy may not emit anything with enough energy to trigger Cerenkov radiation.

Cerenkov radiation can occur in most scintillating fibers. In the prior art, there are two common ways to remove the errors arising from Cerenkov radiation. The first is to employ special filter assemblies within the measuring fiber to remove the Cerenkov light component (see e.g. FIG. 1 in JP 2001-56381 A). The second is to include an additional reference (or background) fiber, without a scintillating tip, which is exposed to the ambient radiation (see FIG. 1 in Phys. Med. Biol., 1992, Vol. 37 No. 10, at page 1886). The reference fiber also produces the Cerenkov light component which can then be subtracted from the output signal of the measuring fiber to produce a corrected signal substantially free of the Cerenkov light component.

A third way to remove the Cerenkov light component is mathematically. This way is preferred since physical modifications

to the fiber assembly are not required and it can therefore lead to a simpler and cheaper way to perform in-vivo radiation dosimetry. However, previous attempts to mathematically model or predict the Cerenkov radiation component have required complex software to evaluate complex mathematical formulas and have also required a knowledge of the geometrical shape of the fiber path (see FIGS. 12 and 13, and sections [0062] to [0075], of JP 2001-56381 A).

Summary of the Invention

It is a first object of the invention to provide high-resolution 1-dimensional, in-vivo dosimetry with a single, scintillating fiber. This fiber can be very small in diameter (e.g. 0.5 mm). This allows radiation measurement along any path in the human body that a 1 mm diameter needle (the standard blood sample needle size) or catheter can go.

It is a second object of the invention to provide automatic removal of the error from Cerenkov radiation in the use of scintillating fibers, without needing a knowledge of the geometrical shape of the fiber path or complex software, thereby providing a simpler and cheaper way to perform in-vivo dosimetry.

In one respect, the invention relates to a displacement difference dosimetry method for in-vivo measuring of dose rates within a radiation field, the method comprising the steps of:

a) providing a scintillating fiber having an insertion end and a coupling end;

b) coupling the coupling end of the scintillating fiber to a

light intensity measuring device, the light intensity measuring device being located substantially outside of the radiation field and producing a voltage output in accordance with a measured light intensity from the scintillating fiber;

c) providing a guide channel having an insertion end and an external end;

d) inserting the insertion end of the guide channel into a human body to a region where radiation is to be measured, so as to provide a substantially fixed path into the human body;

e) inserting the insertion end of the scintillating fiber into the external end of the guide channel and into the human body along the substantially fixed path;

f) subjecting the region of the body at the insertion end of the scintillating fiber to radiation;

g) detecting the position ℓ of the insertion end of the scintillating fiber along the substantially fixed path within the human body and measuring the light intensity at the light intensity measuring device, with the measured light intensity representing both scintillation light from the scintillating fiber and also Cerenkov light;

h) incrementally displacing the insertion end of the scintillating fiber by a small distance $\Delta\ell$ to a new detected position $\ell+\Delta\ell$ along the substantially fixed path and measuring the light intensity at the light intensity measuring device, with the measured light intensity representing both scintillation light from the scintillating fiber and also Cerenkov light;

i) determining a radiation dose rate, substantially free from

the effects of Cerenkov light, for an incremental segment from ℓ to $\ell+\Delta\ell$ along the substantially fixed path according to the expression:

$$\text{Dose Rate} = C \times \Delta V/\Delta\ell,$$

where C is a coefficient, ΔV is the change in voltage output of the light intensity measuring device which results from the insertion end of the scintillating fiber being moved between the positions ℓ and $\ell+\Delta\ell$, and $\Delta\ell$ is the amount of incremental displacement.

In another respect the invention relates to a displacement difference dosimetry method as described above wherein the voltage output produced by the light intensity measuring device varies substantially linearly in accordance with the dose rate of radiation hitting the scintillating fiber.

In still another respect the invention relates to a displacement difference dosimetry method for in-vivo measuring of dose rates within a radiation field, the method comprising the steps of:

a) providing a scintillating fiber having an insertion end and a coupling end;

b) coupling the coupling end of the scintillating fiber to a light intensity measuring device, the light intensity measuring device being located substantially outside of the radiation field and producing a voltage output which varies substantially linearly in accordance with a dose rate of radiation hitting the scintillating fiber;

c) providing a guide channel having an insertion end and an external end;

d) inserting the insertion end of the guide channel into a human body to a region where radiation is to be measured, so as to provide a substantially fixed path into the human body;

e) inserting the insertion end of the scintillating fiber into the external end of the guide channel and into the human body along the substantially fixed path;

f) subjecting the region of the body at the insertion end of the scintillating fiber to radiation;

g) detecting the position ℓ of the insertion end of the scintillating fiber along the substantially fixed path within the human body and measuring the light intensity at the light intensity measuring device;

h) incrementally displacing the insertion end of the scintillating fiber by a small distance $\Delta\ell$ to a new detected position $\ell+\Delta\ell$ along the substantially fixed path and measuring the light intensity at the light intensity measuring device;

i) determining a radiation dose rate for an incremental segment from ℓ to $\ell+\Delta\ell$ along the substantially fixed path according to the expression:

$$\text{Dose Rate} = C \times \Delta V/\Delta\ell,$$

where C is a coefficient, ΔV is the change in voltage output of the light intensity measuring device which results from the insertion end of the scintillating fiber being moved between the positions ℓ and $\ell+\Delta\ell$, and $\Delta\ell$ is the amount of incremental displacement; and

after the radiation dose rate for the incremental segment from ℓ to $\ell+\Delta\ell$ along the substantially fixed path has been determined, repeatingly performing the following additional step to provide 1-

dimensional radiation tomography:

j) incrementally displacing the insertion end of the scintillating fiber by an additional small distance, detecting the resulting change in position of the insertion end of the scintillating fiber and measuring the corresponding light intensity from the scintillating fiber at the light intensity measuring device, and determining a new radiation dose rate for an additional incremental segment along the substantially fixed path according to the Dose Rate expression.

In each case, the coefficient C may be equal to 1 (thus yielding a relative Dose Rate expression) or may be a derived calibration coefficient (thus yielding an absolute Dose Rate expression). The guide channel may be a catheter, hypodermic needle, or similar device.

The invention will, however, be best understood by a perusal of the following description in conjunction with the accompanying drawing, in which:

Brief Description of the Drawings

FIG. 1 is a schematic representation of an in-vivo dosimetry method according to the preferred embodiments of the invention using a single scintillating fiber inserted into the human body.

Detailed Description of the Preferred Embodiments

Referring now to FIG. 1, there is shown schematically a radiation source 1 producing a radiation field 2 within a human body, the outline of which is indicated at 3. The radiation source 1 may comprise an internal or external High Dose Rate source (such as are used in HDR afterloader therapy or external beam therapy), an internal Low Dose Rate source (such as an implanted brachytherapy seed or radio-pharmaceuticals), or any other radiation source used in treating the human body. A guide channel 4 (such as a catheter or hypodermic needle having an internal diameter of between 0.3 and 1.1 mm and an external diameter of between 0.4 mm and 1.5 mm, shown only partially in FIG. 1) is inserted into the human body 3 and passes in the vicinity of the radiation field 2. The guide channel 4 (which provides very little or no radiation shielding) has an insertion end 4a (which is inserted into the human body) and an external end 4b (which remains outside of the human body). The position of the insertion end 4a of the guide channel 4 within the human body 3 can be determined using standard X-ray or ultrasound techniques, and the guide channel 4 provides a fixed (or substantially fixed) path into the human body 3. A single, continuous, flexible scintillating fiber 5 having a sub-mm diameter and one or more cladding layers (for preventing, or greatly reducing, the loss of optical light within the fiber to its surroundings) is inserted into the guide channel 4 and also passes into the human body in the vicinity of the radiation field 2. The scintillating fiber 5 (of conventional

composition and having a length of between 0.25 and 2.0 meters and a diameter of between 0.25 and 1.0 mm, with 0.5 mm being preferred) has an insertion end 5a (or tip, which is inserted into the human body) and an external end 5b (which is shown only schematically in FIG. 1 and which remains outside of the human body, being shielded from ambient light e.g. by means of an extramural absorber coating on the scintillating fiber wall, or by encasing the scintillating fiber in a thin wall, opaque, polymer tubing). The external end 5b of the scintillating fiber is optically (and in the preferred embodiments, mechanically) coupled to a light intensity measuring device 6. The light intensity measuring device 6, which in the preferred embodiments comprises a photo-multiplier tube (PMT), is located substantially outside of the radiation field 2 and produces a voltage output 6a in accordance with the magnitude of light intensity emanating from the external end 5b of the scintillating fiber 5. In the preferred embodiments of the invention, the light intensity measuring device 6 responds substantially linearly to incident light and produces a voltage output 6a which is linearly related to, or substantially linearly related to, the dose rate of radiation hitting the scintillating fiber. (The dose rate is proportional to the number of radiation emissions per second.) Semiconductor diode modules may also be employed as the light intensity measuring device 6. Displacing means 7 are provided for inserting the scintillating fiber 5 into and withdrawing the scintillating fiber 5 from the guide channel 4 (and human body) by pushing or pulling. The displacing means 7 may be mechanical (such as a motor coupled to a wheel which linearly drives the

scintillating fiber 5) or manual (that is, the movement of the scintillating fiber 5 can be imparted by the hand of an operator). Displacement detecting means 8 are provided for detecting the amount of displacement of the scintillating fiber 5 imparted by the displacing means 7 and producing an output signal 8a representative thereof. The displacement detecting means 8 may comprise any device which is convenient (such as a rotary encoder coupled to a wheel which is driven by the linear movement of the scintillating fiber 5), and is employed to provide information about both the incremental and the absolute position of the internal end 5a of the scintillating fiber 5 within the guide channel (and human body).

The voltage output 6a from the light intensity measuring device 6 and the output signal 8a from the displacement detecting means 8 are fed to a processing circuit 9, the function and operation of which is described below. The processing circuit 9 includes an additional input 10 which receives information about the exact 3-dimensional position of the tip of the scintillating fiber within the human body, e.g. relative to a body organ or body region to be irradiated, or relative to the guide channel 4. The input signal 10 is produced by standard medical imaging means (e.g. X-ray or ultrasound) used to locate the tip of the scintillating fiber 5 before and/or at various times during the radiation measurement procedure. The processing circuit 9 also includes an optional but useful calibration input 11, the purpose of which will be described in greater detail below.

The radiation dosimetry method according to a first preferred embodiment of the invention includes the following steps:

• inserting the insertion end 4a of the guide channel 4 into a human body 3 so as to provide a substantially fixed path into the human body 3;

• inserting the insertion end 5a of the scintillating fiber 5 into the external end 4b of the guide channel and into the human body 3 along the substantially fixed path;

• subjecting the insertion end 5a of the scintillating fiber to the radiation field 2;

• using the displacement detecting means 8 and/or the input signal 10 from the standard medical imaging means, detecting an initial position ℓ ("L" in FIG. 1, as indicated by the phantom line) of the insertion end 5a of the scintillating fiber 5 along the substantially fixed path defined by the guide channel 4 within the human body 3 and sending an initial position signal to the processing circuit 9; and using the light intensity measuring device 6, detecting the light intensity emanating from the external end 5b of the scintillating fiber 5, and sending the voltage output 6a (representing both scintillation light from the scintillating fiber 5 and also Cerenkov light) corresponding to the initial position of the insertion end 5a of the scintillating fiber 5 to the processing circuit 9;

• storing the initial position signal and its corresponding voltage output 6a in the processing circuit 9;

• using the displacing means 7, incrementally displacing (either step-wise or continuously) the insertion end 5a of the scintillating fiber 5 by a small distance $\Delta\ell$ ("delta L" in FIG. 1) to a new position $\ell+\Delta\ell$ along the substantially fixed path;

• using the displacement detecting means 8, detecting the change in position $\Delta\ell$ of the insertion end 5a of the scintillating fiber 5 along the substantially fixed path and sending a change in position signal to the processing circuit 9 corresponding to the new position; and using the light intensity measuring device 6, detecting the light intensity emanating from the external end 5b of the scintillating fiber 5, and sending the voltage output 6a (representing both scintillation light from the scintillating fiber 5 and also Cerenkov light) corresponding to the new position of the insertion end 5a of the scintillating fiber 5 to the processing circuit 9;

• storing the change in position signal and its corresponding voltage output 6a in the processing circuit 9;

• in the processing circuit 9, determining a radiation dose rate, substantially free from the effects of Cerenkov light, for an incremental position of the insertion end (or tip) 5a of the scintillating fiber 5 as the insertion end 5a occupies a segment from ℓ to $\ell+\Delta\ell$ along the substantially fixed path defined by the guide channel 4 according to the Dose Rate equation:

$$\text{Dose Rate} = C \times \Delta V/\Delta\ell, \qquad (1)$$

where C is a calibration coefficient (optional), ΔV is the change in voltage output of the light intensity measuring device which results from the insertion end 5a of the scintillating fiber 5 being moved between the initial and new positions ℓ and $\ell+\Delta\ell$, and $\Delta\ell$ is the amount of change in position of the insertion end 5a of the scintillating fiber; and

• repeatingly incrementally displacing the insertion end 5a of

the scintillating fiber 5 by a small distance, detecting the change in position of the insertion end 5a of the scintillating fiber 5 and the corresponding light intensity emanating from the scintillating fiber, and determining the corresponding radiation dose rate for each incremental position of the insertion end 5a of the scintillating fiber 5 (i.e. as the insertion end 5a occupies additional incremental segments along the substantially fixed insertion path defined by the guide channel 4) according to the Dose Rate equation (1).

The radiation dosimetry method according to a second preferred embodiment of the invention includes the following steps for providing high-resolution imaging:

- inserting the insertion end 4a of the guide channel 4 into a human body 3 so as to provide a substantially fixed path into the human body 3;
- inserting the insertion end 5a of the scintillating fiber 5 into the external end 4b of the guide channel and into the human body 3 along the substantially fixed path;
- subjecting the insertion end 5a of the scintillating fiber to the radiation field 2;
- using the displacement detecting means 8 and/or the input signal 10 from the standard medical imaging means, detecting an initial position ℓ ("L" in FIG. 1, as indicated by the phantom line) of the insertion end 5a of the scintillating fiber 5 along the substantially fixed path defined by the guide channel 4 within the human body 3 and sending an initial position signal to the processing circuit 9; and using the light intensity measuring

device 6, detecting the light intensity emanating from the external end 5b of the scintillating fiber 5, and sending the voltage output 6a corresponding to the initial position of the insertion end 5a of the scintillating fiber 5 to the processing circuit 9;

• storing the initial position signal and its corresponding voltage output 6a in the processing circuit 9;

• using the displacing means 7, incrementally displacing (either step-wise or continuously) the insertion end 5a of the scintillating fiber 5 by a small distance $\Delta\ell$ ("delta L" in FIG. 1) to a new position $\ell+\Delta\ell$ along the substantially fixed path;

• using the displacement detecting means 8, detecting the change in position $\Delta\ell$ of the insertion end 5a of the scintillating fiber 5 along the substantially fixed path and sending a change in position signal to the processing circuit 9 corresponding to the new position; and using the light intensity measuring device 6, detecting the light intensity emanating from the external end 5b of the scintillating fiber 5, and sending the voltage output 6a corresponding to the new position of the insertion end 5a of the scintillating fiber 5 to the processing circuit 9;

• storing the change in position signal and its corresponding voltage output 6a in the processing circuit 9;

• in the processing circuit 9, determining a radiation dose rate for an incremental position of the insertion end (or tip) 5a of the scintillating fiber 5 from ℓ to $\ell+\Delta\ell$ as the insertion end 5a occupies a segment from ℓ to $\ell+\Delta\ell$ along the substantially fixed path defined by the guide channel 4 according to the Dose Rate equation:

$$\text{Dose Rate} = C \times \Delta V/\Delta \ell, \qquad (1)$$

where C is a calibration coefficient (optional), ΔV is the change in voltage output of the light intensity measuring device which results from the insertion end 5a of the scintillating fiber 5 being moved between the initial and new positions ℓ and $\ell+\Delta\ell$, and $\Delta\ell$ is the amount of change in position of the insertion end 5a of the scintillating fiber; and

• repeatingly incrementally displacing the insertion end 5a of the scintillating fiber 5 by a small distance, detecting the change in position of the insertion end 5a of the scintillating fiber 5 and the corresponding light intensity emanating from the scintillating fiber, and determining the corresponding radiation dose rate for each incremental position of the insertion end 5a of the scintillating fiber 5 (i.e. as the insertion end 5a occupies additional incremental segments along the substantially fixed insertion path defined by the guide channel 4) according to the Dose Rate equation (1).

It will be seen that the above methods provides means to achieve 1-dimensional radiation tomography along an arbitrary (but fixed) curved or straight path. Simple arguments show that the ratio of the change in the external light intensity (measured at the PMT) divided by the displacement of the fiber is proportional to the dose rate at the internal scintillating fiber tip, if the fiber path in the region of the radiation field does not change. Additionally, the derived Dose Rate(s) is/are substantially free from the adverse effects of Cerenkov light (as well as all contributions which are substantially constant during the

displacement $\Delta\ell$). This is shown in the following scenario.

Imagine a scintillating fiber having some (unknown but fixed) path within the body, and extending outside the body to a means of measuring light intensity. A measurement of the light (due to fiber irradiation) in the scintillating fiber arriving at the light measuring device is made. Now imagine a small segment of length, $\Delta\ell$, of the scintillating fiber being cut off at the inner tip of the fiber and attached at the outer tip of the fiber which is at the input of the light intensity measuring device, and where the radiation intensity is negligible. A second measurement of light intensity is now made. The difference of the two light intensity measurements, which results from the light generated in '$\Delta\ell$', is called 'ΔI'. If the light measuring device responds linearly (or substantially linearly) to the incident light, then the Dose Rate equation, given above at (1), results. (A small correction for attenuation of light in the scintillating fiber that results in this process is omitted from the Dose Rate equation. This correction is ~ 1% (negligible), but can be easily included in the Dose Rate equation. All other contributions to the measured light intensity, which complicate the dosimetry measurement, are removed with this difference measurement. This cancels out, for example, most (~99%) of the Cerenkov radiation generated along the fiber remaining in the body. Only radiation light from the fiber tip survives the difference measurement. Here isotropic scintillation light will greatly outweigh the Cerenkov light.) The Dose Rate equation, with no derived knowledge of the calibration coefficient C (i.e., with no further measurements, and using C=1 in the Dose

Rate equation and for the calibration input 11 to the processing circuit 9 in FIG. 1), yields a relative Dose Rate equation useful in finding edges and other features (e.g. local maximums) of radiation fields. On the other hand, if the voltage output of the light intensity measuring device is linearly (or substantially linearly) related to the dose rate of the radiation hitting the scintillating fiber, then C is a constant, and a single calibration measurement will yield the value of C, permitting an absolute Dose Rate to be obtained in all further measurements. This allows measurement of in-vivo radiation without complications from Cerenkov (and other in-fiber) contributions.

In order to obtain absolute Dose Rate values by the above radiation dosimetry method, only one calibration run (to find C in the Dose Rate equation) is usually necessary. This calibration can be carried out in a safe (dummy) material and in an environment where the dose rate is known and where ΔV and $\Delta \ell$ can be sensed or detected. Solving the Dose Rate equation for C (and the use the derived calibration coefficient C for the calibration input 11 to the processing circuit 9 in FIG. 1) is then a straight-forward process.

In the case where the Dose Rate equation is used with a 0.5 mm diameter scintillating fiber and a photo-multiplier tube (PMT) having a gain of $8.5*10^{10}$ V/W, and where the incremental displacement of the scintillating fiber tip 5a is 0.5 mm, then a ΔV at the PMT output of approximately 2.5 mV will result when the scintillating fiber tip is in a region of a 1 mGray/sec dose rate. This can be measured in approximately 1 second (to provide for

averaging with signal integration) or less. If desired, the scintillating fiber can be moved at constant speed and data can be taken periodically; dosimetry can thereby be achieved along the fiber path from the ΔV signals produced from the PMT output. Measurements of 10 cm paths with sub-mm resolution will take only a few seconds.

This low-cost dosimetry method, usable with small, battery-operated PMTs, has simplicity, smallness, and sufficient responsivity for almost all "in-vivo" measurements. Data logging, computer storage, & fast radiation control are easily achieved. Uniformity of fiber radiation response can be independently measured, if necessary. The use of endoscopes and radio-pharmaceuticals can provide special features and performance benefits in this dosimetry.

While the above invention has been described with certain particularity, it is not meant to be limited to the above described preferred embodiments. Therefore, the invention will encompass the preferred embodiments described above as well as any modifications thereof which will fall within the scope of the appended claims.